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Additive Manufacturing Technologies Used for 3D Metal Printing in Dentistry

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Abstract

Purpose of Review: Compared to conventional casting methods used for processing different alloys for dental applications, additive manufacturing technologies reduce manufacturing time and costs, minimize human errors and prevent possible defects in the cast objects. This review highlights working mechanisms, possible advantages and drawbacks of recent additive manufacturing technologies used for metal processing in dentistry. **Recent Findings:** The literature reviewed indicated that powder based fusion mainly based on selective laser sintering, selective laser melting and electro beam melting are the most commonly used technologies for 3D metal printing in dentistry for dental appliances made of CoCr and Ti6Al4V. Although mechanical properties of 3D printed alloys could be considered satisfactory, accuracy and reproducibility data do not present consistent results. **Summary:** There appears room for improvement between 3D printed metals and ceramic interfaces and precision before such technologies could be favoured over conventional cast methods.

Keywords 3D printing ● Additive Manufacturing ● Electron Beam Melting ● Metal ● Prosthodontics ● Selective Laser Sintering ● Selective Laser Melting

Introduction

Technologies based on Computer Aided Design and Computer Aided Manufacturing (CAD-CAM) typically comprise three fundamental elements embedded in the digital workflow: (1) data acquisition or digitizing, (2) data processing (CAD) and (3) manufacturing (CAM) [1]. The most commonly used CAM process today is the computer numerically controlled (CNC) machining that is based on routes where power-driven machinery tools, such as saws, lathes, milling machines, and sharp cutting drills or tools that mechanically cut the material to achieve the desired geometry with all the steps controlled by a computer program [2-4]. However, such technologies present a number of manufacturing limitations, namely, a considerable amount of raw material is wasted as unused parts of the milled blocks, the milling tools needs to be replaced after a short number of running cycles due to the heavy abrasion and wear when milling and the size of the milling burs and the axis of the CNC machine may limit the access to the small areas of the block to be milled [5-8].

As an alternative to the subtractive methods, additive processes provide manufacturing procedures in which the powder or liquid base material is build into a solid object [9,10]. The American Society for Testing and Materials (ASTM) has defined Additive Manufacturing (AM) as ‘‘a process of joining materials to make objects from 3D model data, usually layer upon layer, as opposed to subtractive manufacturing methodologies” [11]. The CAD data files exported for the industry-standard exchange file format is the Standard Triangulation Language (STL) that is a boundary representation consisting a list of triangular facets [12]. In 2008, the ASTM international committee F42 on AM technologies has classified seven categories: stereolithography (SLA), material jetting (MJ), material extrusion (ME), binder jetting (BJ), powder based fusion (PBF), sheet lamination (SL) and direct energy deposition (DEP) [11].

Powder based fusion technologies

Powder based fusion (PBF) is the most commonly used technology for 3D metal printing in dentistry. Currently, three types of PBF technologies are available, namely selective laser sintering (SLS), selective laser melting (SLM) and electro beam melting (EBM) [9,11]:

Selective Laser Sintering (SLS)

Selective Laser Sintering (SLS) technology was developed by Carl Deckard and Joe Beaman and patented in 1989 [13,14] that is based on application of high powered laser (Na:YAG laser) beam focused onto a bed of powdered metal where thin solid layers (20-100 μm) are fused until the 3-dimensional (3D) object is built [15]. The entire fabrication chamber is sealed and maintained at a temperature just below the melting point that the metal powder sinters [16]. Object parts made by partial melting are characterized by high porosity with initially achieving only point contacts between the particles. During laser heating, various sintering and rearrangement mechanisms induce the powder binding and densification. Unfortunately, with partial melting and sintering mechanism, porosities could not be complete eliminated as generally possible repulsion forces arise between particles at a high fraction of the binding liquid component [17].

Selective Laser Melting (SLM)

With the introduction of powerful high-quality lasers, the partial melting achieved by SLS technology has been taken over by complete melting, giving rise to a new development of metal laser sintering (MLS) or SLM [18-21]. Although the superficial finish is considered very well, the components may have high internal stresses caused by thermal gradients induced during processing and therefore require additional heat treatment [20]. The most common fibre laser used on the SLM technology is the CO_2 laser (1-2 kW) for processing metal powders and the building plate can be preheated up to 200ºC [21].

Electron Beam Melting (EBM)

Instead of using a laser beam (60kW) to melt or sinter the powder, with the EBM technologies, a focused electron beam is used to selectively melt layers of powder (100 μm) in an inert environment such as purified argon. Also, while building the part, an elevated temperature of about 700ºC is maintained in the chamber to reduce the residual stresses. Initially, a tungsten filament is heated over 3000ºC, which causes electrons to be emitted, and subsequently, potential difference between a cathode and an anode causes the electrons to accelerate. The electrons are focused and detected using magnetic coils to form a narrow high-energy beam that attacks the surface of the powder. Eventually, the kinetic energy transferred through friction creates the heat that is necessary to melt the metal powder [19,22].

The main differences between the PBF technologies are the operational parameters such as melting temperature, energy source, energy power, laser beam absorption/reflections coefficients, thermal conductivity, chamber conditions, temperature reached, along with other parameters such layer thickness, build orientation and grain size [23-25]. All these parameters have to be adjusted depending on the metal type.

Representative examples of implant frameworks using SLM technology is presented in Figs. 1a-d.

Metals used in additive manufacturing in dentistry

For 3D metal printing in dental applications, currently Cobalt-Chrome (Co-Cr) and Titanium (Ti) are the most commonly used alloys.

The metal powder of Co-Cr also contains Molybdenum, Tungsten, Silicon, Cerium, Iron, Manganese and Carbon, while Nickel and Beryllium are not present in the composition anymore. The metal powders used in conjunction with AM technologies are a mixture of particles with sizes ranging between 3 and 14 μm [25,26]. Depending on the manufacturers of the AM technologies, composition (Table 1) and mechanical properties of Co-Cr may show differences between products (Table 2). Likewise, typical titanium alloys used in dentistry and medicine such as Ti6Al4V show slight differences in composition (Table 3) and physical and mechanical properties depending on the AM system (i.e. SLM vs. EBM) (Table 4) [27-30].

Mechanical properties of 3D printed metals

The recommended mechanical properties for the CoCr alloys for fixed and removable dental restorations and appliances are reflected in the ISO 22674 [31]. Few studies compared the mechanical properties of cast, milled and AM alloys used for dental purposes [32-34]. In 2014, Al Jabbari et al [32] evaluated the hardness and microstructural characterization of the CoCr dental alloys manufactured using casting, milling or SLM techniques where significant differences were noted in hardness values being the highest for SLM (371±10 HV), followed by the cast (320±12 HV) and milled (297±5 HV) procedures. The manufacturing process showed also a significant effect on the alloy microstructure where X-ray radiography revealed the presence of porosity only in the cast group. It has to be noted that, although effort was made to use one type of CoCr alloy for all procedures, for SLM manufacturing another type of CoCr alloy had to be used which has slightly higher Co/Cr ratio than those of cast and milled ones. Building direction and tensile directions (0, 45 and 90º) also significantly affected the mechanical properties of CoCr SLM specimens [35] where zero angle build up (building and tensile direction are parallel) demonstrated the highest tensile strengths and elongation after fracture.

The objects made of CoCr through the SLS or SLM technologies present thermal stresses in the body of the object due to rapid heating and cooling during the fabrication process [35]. Such internal residual stress could generate high strain and thereby affects the accuracy [36,37]. One solution to circumvent this problem is the employment of heat treatment after printing the metal that could at the same time change the micro-structure [34,38]. Nevertheless, the reduction in porosity in CoCr alloys fabricated through AM technologies, has a positive effect on the mechanical properties of the printed object [34] through which higher yield and tensile strength of the cast alloys could be obtained [39]. Corrosion resistance of CoCr SLM specimens on the other hand, appears to remain similar to that of the conventional casting processing [40-42].

Ti6Al4V is a two-phase material, consisting of the hexagonal close packed (hcp) α phase and the body center cubic (bcc) β phase [43]. The transition temperature between the two phases for Ti6Al4V is 995°C [44,45]. The mechanical properties of the two-phase Ti6Al4V alloy are dependent on the microstructure and the distribution of the two phases throughout the material [44,46,47]. The SLM technology is more commonly used to manufacture Ti6Al4V fixed dental prostheses (FDPs) than EBM. SLM produces more rapid cooling, resulting in transformation to α′martensite phase in various proportions that significantly affects the corrosion potential and be detrimental for dental applications [43]. However, when low oxygen containing powder is used in EBM fabrication, mechanical properties could be improved. [30]. Mechanical properties of the Ti6Al4V alloy fabricated using EBM versus casting was reported to deliver strength (1.18 GPa) and elongation (16 to 25%) with microindentation hardness ranging from 3.6 to 3.9 GPa, comparable to the very best wrought Ti6Al4V alloy (4 GPa) [43].

Compared to conventional casting method, AM technologies offer the advantages of high product density, reduced manufacturing time and costs, minimization of human errors, and the prevention of casting defects. However, different dental applications would necessitate some prerequisites from AM technologies for optimum outcome.

Dental applications using additive technologies

Removable Partial Dentures (RPD) and overdentures

In 2004, Williams et al [48] described digital surveying a definitive cast, CAD design and SLA AM resin pattern fabrication for casting the RPD metal frameworks. Subsequently, in 2006, Williams et al [49,50] developed the technique for the Co-Cr SLM AM technology to manufacture RPD frameworks. Thereafter, Kattadiyil et al [51] closed the digital workflow incorporating the intraoral digital impression (Cadent Itero, San Jose, CA, USA) of the partially edentulous maxillary arch (Kennedy Class III) for the fabrication of an RPD. The case report described the digital impression where 28 scans were needed for the maxillary teeth and occlusion, 25 scans to enhance the capture of the rest seats and 28 scans for the mandibular teeth yielding to a total of 17 minutes. This remained a major drawback until 2014 when Kanazawa et al, reported a protocol for Ti6Al4V framework for a complete maxillary denture using the SLM technology, with a layer thickness of 30 μm [52]. The digitalization of the definitive plaster cast was executed with a 3D cone-beam computed tomography and the digital design was completed with a non-dental CAD software (Freeform, Geomatic) which made the whole process much more practical [53]. In 2016, Lee et al [54] measured the internal fit of 10 RPDs made for 10 patients using SLM technology. Although no significant differences were found among participants in terms of internal discrepancy of the various framework components, unfortunately the results were not compared to the conventionally fabricated RPDs. Likewise, technique description for a SLM CoCr superstructure framework of a maxillary implant-retained overdenture is available but systematic measurements are lacking [55-62]. AM technologies could be also used to manufacture implant-borne fixed prostheses but due to the roughness and texture of the metals, this technology still needs to be combined with subtractive technologies.

Fixed Dental Prostheses (FDP)

AM technologies were also introduced for the fabrication of CoCr alloy frameworks for crowns and FDPs but the major focus of research is on precision and ceramic adhesion that are critical for the longevity of such reconstructions.

Marginal and internal gap

Marginal and internal gap of CoCr 3-unit FDPs manufactured with conventional lost wax, milling or DMLS fabrication methods indicated the best fit with DMLS group (84±6 μm), followed by the lost wax (133±9 μm) and milled group (166±2 μm) [63]. However, only vertical gaps were measured and no measurements were performed in horizontal planes. In another study, where internal gaps of pre-sintered milled, cast and SLS manufactured CoCr metal frameworks indicated the lowest gap formation with the milled one (32±5 μm), followed by SLS (47.3±9 μm) and casting (64.1±14 μm) [64]. Nevertheless, in this study, absolute marginal discrepancy was not measured which is in fact more of a clinical concern. In a further study, internal gaps of 60 CoCr metal molar crowns fabricated with casting, milling or DMLS in 42 patients were measured [65]. Although overall, no statistical significant differences were found between the three systems, internal gap at the occlusal and axio-occlusal region were higher for DMLS crowns (290.1±112 μm and 188.1±69 μm, respectively) than for the milled (265.7±90 μm and 141.1±53 μm, respectively) and cast (201.1±67 μm and 140.6±48 μm) groups. Huang et al [66] evaluated 330 single-unit CoCr metal ceramic crowns in 274 patients. Each crown was randomly assigned to one of the three groups: CoCr SLM, CoCr cast and AuPt cast. The SLM Co-Cr crowns demonstrated a similar marginal fit (75.6±32.6 μm) to that of the cast Au-Pt ones (76.8±32.1 μm) and a better marginal fit than that of the cast Co-Cr crowns (91±36.3 μm). For the internal gap however, at the occlusal region, the SLM group (309.8±106.6 μm) was less accurate than the CoCr (254.6±109.6 μm) and AuPt (249.6±110.4 μm) cast groups.

Ceramic-metal adhesion

When three-point flexural strength test results are considered, ceramic adhesion to 3D printed CoCr metal alloys exceeded the minimum prerequisite of 25 MPa [65-70] required by the AINSA/ADA specification No. 38 (2000) [71] and ISO 9693:1999(E) [72]. However, although adhesion results seem to be favorable, failure types were mainly adhesive when SLS technology was used whereas conventional cast CoCr and NiCr alloys showed frequently mixed mode of cohesive failures [67-69]. This is often related to the oxide layer morphology on the surface [73], which seems to be less favorable for SLM but area fraction of adherence porcelain could be optimized with 5 to 7 times of multiple firings [74] which is less efficient but seems to increase the marginal adaptation of SLM CoCr reconstructions [75].

Concluding remarks

Additive manufacturing technologies reduce manufacturing time and costs, minimize human errors, and prevent possible defects in the cast objects compared to conventional casting methods applied for dental alloys. Currently, such technologies are more commonly used for processing CoCr and Titanium for dental applications. Among different technologies, available data in the dental literature accentuated more on Selective Laser Melting and Selective Laser Sintering. However, accuracy and reproducibility data do not present consist results and there appears room for improvement at 3D printed metals and ceramic interfaces and oxide layer morphologies to favour such technologies over conventional cast methods.

Conflict of Interest

The authors declare that they have no conflict of interest.

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Table 1. Brands and chemical composition of CoCr alloys provided for additive manufacturing.

Brand	**Chemical Composition (wt%)**	
EOS CoCr MP1	Co:60-65 Cr: 26-30 Mo: 5-7 Si≤1	Fe≤0.75 Mn≤1 C≤0.16 Ni≤0.1
EOS CoCr SP2	Co:63.8 Cr: 24.7 Mo: 5.1 W: 5.4 Si: 1	Fe<0.75 Mn<0.1
Renishaw CoCr DG1	Co:63.9 Cr: 24.7 Mo: 5 Ni<0.5 N<0.25	Fe<0.5 Mn<1 Al<0.10 0<0.10 C<0.05
SLM Solutions **CoCr28Mo6 acc to ASTM F75**	Co: Balance Cr: 27-30 Mo: 5-7 W: 0.20 Si: 1	Al: 0.10 Fe: 0.75 Mn: 1 C: 0.35 N: 0.25 B, S: 0.01
SLM Solutions **MediDent**	Co: Balance Cr: 22.7-26.7 Mo: 4-6 W: 4.4-6.4 Si: 2	Fe: 0.5 Mn: 0.10 C: 0.02 Ni: 0.10 B, S: 0.10
3D systems LayerWise **CoCr ASTM F75**	Co: Balance Cr: 27-30 Mo: 5-7 W≤0.2 Si≤1	Fe≤0.75 Mn≤1 C≤0.35 Ni≤0.5 B,S≤0.01 P≤0.02 Al, Ti≤0.1 N≤0.25
3D Systems LayerWise **CoCr 3DS Dentwise**	Co: 59 Cr: 25 W: 9.5 Mo: 3.5 Si: 1	C, Fe, Mn, N: <1.5
Concept Laser **Remanium star CL**	Co:60.5 Cr: 28 W: 9 Si: 1.5	Fe, Mn N, Nb and free form Ni, Be, Ga<1
BEGO	Co: 63.9 Cr: 24.7 W: 5.4 Mo: 5.0 Si≤1	

Property	**EOS MP1**	**EOS SP2**	**Renishaw**	**SLM Solutions CoCr28Mo6**	**SLM Solutions MediDent**	**Concept Laser Remanium star CL**	**3D systems ASTM F75**	**3D Systems 3DS Dentwise**	**BEGO**
Alloy type ISO 22674	4	4	4	4	4	5	5	4	5
Density (g/cm^3)	8.3	8.5	8.5	NA	NA	8.6	8.35	8.8	8.5
Tensile strength (MPa)	1100	1350	1097-1104	1101-1039	1062	1030	1000	910	1150-1400
Yield strength (MPa)	600	850	683-714	720-705	319	635	650	650	790-1000
Elongation at break (%)	20	3	16-21	10	NA	10	20	8	9
Young´s modulus (GPa)	200	200	220	194-191	114	230	230	200	210
Hardness (HV)	350-450	420	400-412	375-372	NA	NA	400	310	360
Coefficient thermal Expansion (ºC)	$13.6x10^{-6}$	$14.5x10^{-6}$	$10.2x10^{-6}$	NA	NA	$14.1x10^{-6}$/ºC	$14.3x10^{-6}$/ºC	$14.0x10^{-6}$/ºC	$14.1x10^{-6}$/ºC
Melting interval (ºC)	1350-1430	1410-1450	1260-1482	NA	NA	1320-1420	1350-1430	1305-1400	1370-1420

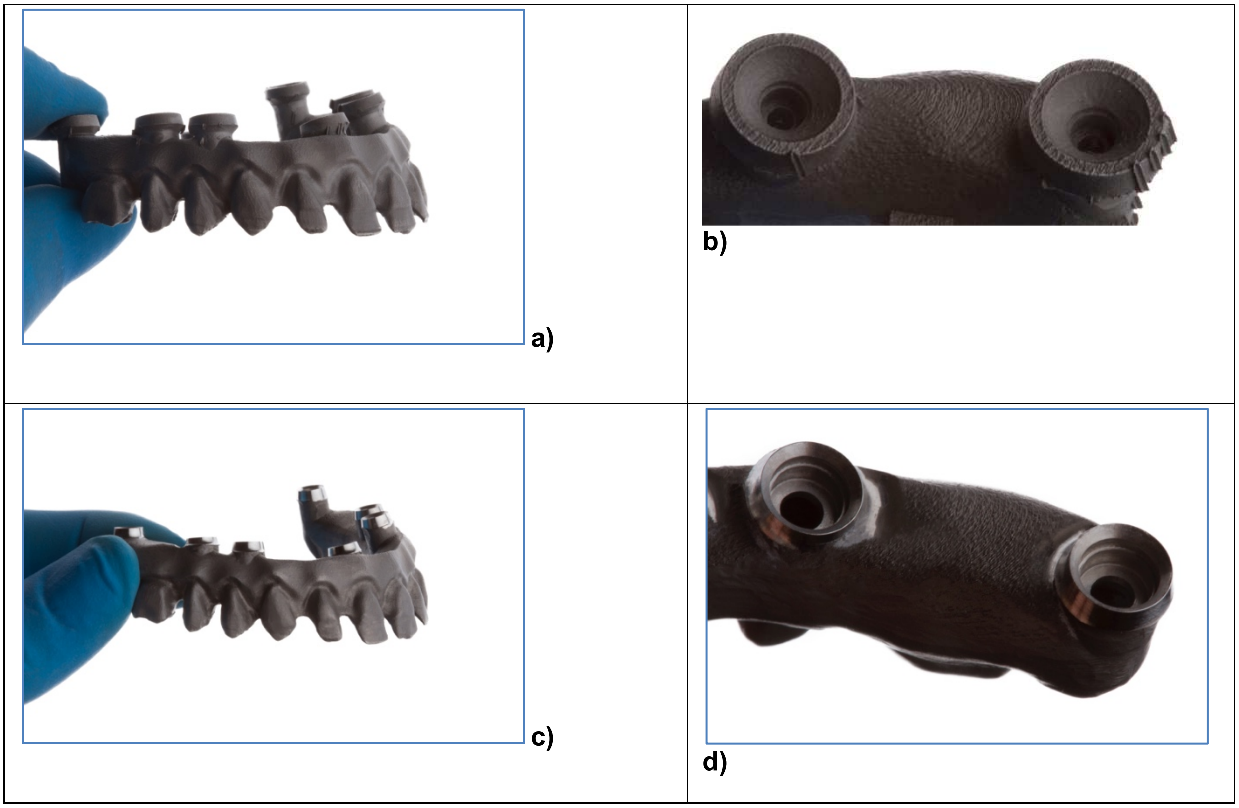
Table 2. Physical and mechanical properties of 3D printed CoCr (after stress relief) provided by their manufacturers.

Brand	Composition (wt%)	
EOS DMLS **Ti64ELI**	Ti: Balance Al: 5.5-6.5 V: 3.5-4.5 O<0.110	N<0.04 C<0.08 H<0.012 Fe<0.250 Y<0.005
Renishaw **Ti6Al4V ELI-0406**	Ti: Balance Al: 5.5-6.75 V: 3.5-4.5 O<0.2	N<0.05 C<0.08 H<0.015 Fe<0.3
SLM Solutions **Ti6Al4V ELI**	Ti: Balance Al: 5.5-6.5 V: 3.5-4.5	Fe: 0.25 C: 0.08 N: 0.03 O: 0.13 H: 0.0125
Concept Laser **Rematitan**	Ti: 90 Al: 6 V: 4	N, C, H, Fe and O <1
3D systems **LaserForm Ti Grade 23**	Ti: Balance Al: 5.5-6.5 V: 3.5-4.5 O≤0.13	N≤0.03 C≤0.08 H≤0.012 Fe≤0.25 Y≤0.005
Arcam EBM **Ti6Al4V ELI**	Ti: Balance Al: 5.5-6-5 V: 3.4-4.5 O<0.13	N<0.05 C<0.08 H<0.012 Fe<0.25

Table 3. Brands and chemical composition of Ti alloys provided for additive manufacturing.

Property	**EOS Ti64ELI**	**Renishaw**	**SLM Solutions Ti6Al4V ELI**	**Concept Laser Rematitan**	**3D systems**	**ARCAM EBM**
Grade/type	NA	23	23	4	23	5
Density (g/cm^3)	4.41	4.42	NA	4.5	4.42	NA
Tensile strength (MPa)	1070	1089	1286	1005	940	860
Yield strength (MPa)	1010	1007	1116	950	850	795
Elongation at break (%)	14	14	8	10	15	10
Young´s modulus (GPa)	NA	129	111	115	105-120	114
Hardness (HV)	34	38	38	NA	30	NA
Coefficient thermal expansion	NA	8-9x10^{-6}/ºC	NA	10.16x10^{-6}/ºC	9.7x10^{-6}/ºC	NA
Melting interval (ºC)	NA	1635-1665	NA	1604-1655	1692-1698	NA

Table 4. Physical and mechanical properties of 3D printed Ti alloys provided by their manufacturers. NA: Not available.



a) b) c) d)

Figs. 1a-d. a) SLM additive manufactured Complete maxillary CoCr implant framework, **b)** detailed photo of texture of the 3D printed framework, **c)** final implant framework and **d)** detailed aspect after the milled implant interface.